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(11) Publication number : **0 654 684 A2**

(12)

EUROPEAN PATENT APPLICATION

(21) Application number : **94308390.7**

(51) Int. Cl.⁶ : **G01T 1/29**

(22) Date of filing : **14.11.94**

(30) Priority : **19.11.93 US 155015**

(43) Date of publication of application :
24.05.95 Bulletin 95/21

(84) Designated Contracting States :
DE FR GB NL

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(54) **Nuclear camera systems.**

(57) A nuclear camera system, e.g. for medical diagnostic imaging, comprises a plurality of radiation detector heads (10a, 10b, 10c) mounted on a rotatable portion (16) of a gantry (18). A transmission radiation source and collimator assembly (40, 42) is mounted on the rotatable gantry portion opposite one of the detector heads. The radiation source (60) of the assembly is mounted in a lead shield (62) with an opening (64) pointing toward an examination region (12). A shutter (66) is rotatable between a closed position in which a lead arc segment (70) blocks the opening (64), a calibration orientation in which tin (72) covers opening (64), and an open position in which an opening (74) is aligned with the opening (64). A safety interlock means (80) locks the shutter in the closed position and against rotation when the radiation source and collimator assembly is removed from the rotatable gantry portion. The collimator (42) has lead side walls (200) and tin or tin alloy septa (202). When the radiation from the radiation source strikes the lead side walls, they emit characteristic lead x-rays of about 88 keV whereas the tin radiates x-rays of only about 30 keV. A filter (210) includes an inner layer (212) of tin for attenuating any remaining 88 keV lead x-rays coming from the transmission line source assembly and converting them to about 30 keV radiation. The filter has an outer layer (214) for attenuating the radiation around 30 keV.

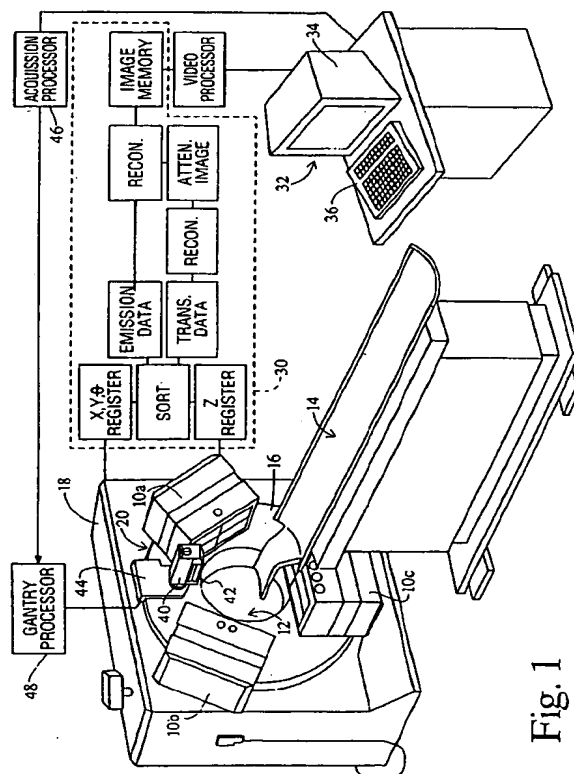


Fig. 1

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The present invention relates to nuclear camera systems, such as are used for medical diagnostic imaging. It finds especial application in conjunction with single-photon emission computed tomography (SPECT) camera systems that include one or more transmission radiation line sources and will be described with particular reference thereto. It is to be appreciated, however, that the invention will also find application in conjunction with other types of nuclear camera system.

Heretofore, single photon emission computed tomography has been used to study the radionuclide distribution in subjects. Typically, one or more radiopharmaceuticals were injected into a patient's blood stream for imaging the circulatory system or specific organs which absorb the injected radiopharmaceuticals. One or more gamma or scintillation camera heads were placed closely adjacent to a surface of the patient to monitor and record radiation emitted by the radiopharmaceuticals. In single photon emission computed tomography, the heads were rotated or indexed around the subject to monitor the emitted radiation from a multiplicity of directions. The data monitored from the multiplicity of directions was reconstructed into a three-dimensional image representation of the radiopharmaceutical distribution within the patient.

One of the problems with the SPECT imaging technique is that the patient is not completely homogeneous in terms of radiation attenuation or scatter. Rather, the human patient includes many different tissue and bone types which absorb or scatter radiation from the radiopharmaceuticals to different degrees. The SPECT images can be made more accurate if they are corrected for the radiation lost to scattering or attenuation along each path through the human torso.

As described in our U.S. Patent No. 5,210,421, a radiation line source can be positioned opposite one or more of the gamma or scintillation camera heads. Transmission radiation from the line source received by the opposite detector head can be reconstructed into a three-dimensional image representation of the radiation absorptive properties of each incremental volume element of the patient, analogous to a CT scan. This radiation attenuation information is utilized to correct the SPECT data. When the radiation line source and the radiopharmaceuticals have distinctly different energy peaks, the transmission radiation and photon emission radiation image data can be collected concurrently and separated based on energy.

In the prior art, collimators were typically constructed of lead. When lead is struck with incident radiation, such as the radiation from the line source, the lead emits an x-ray with a characteristic energy of about 88 keV. The 88 keV x-ray has an energy which is sufficiently close to the emission energy of some common radiopharmaceuticals that it is difficult to

distinguish the two. This inability to distinguish reliably radiation from the radiopharmaceuticals and radiation emitted from the lead caused errors in the resultant radiopharmaceutical image.

One solution for separating the lead x-rays coming from the line source assembly from the radiation emitted from the radiopharmaceuticals was to conduct the line source transmission radiation examination first. The transmission line source was removed or closed before the radiopharmaceuticals were injected and imaged. However, performing the transmission examination and the radiopharmaceutical diagnostic examination sequentially lead to registration problems. The transmission radiation data and radiopharmaceutical images data frequently became at least partially misaligned. This misalignment caused incorrect transmission radiation based corrections on the radiopharmaceutical data causing further image degradation.

The present application provides a nuclear camera system which overcomes the above-described problem.

According to the invention there is provided a nuclear camera system which includes a transmission radiation source which allows primary radiation of a characteristic energy to be transmitted toward a subject, and a collimator for collimating the primary radiation transmitted by the radiation source, at least one of the radiation source and collimator tending to emit secondary radiation in response to incident radiation from the radiation source, characterized in that the system includes means for limiting the effect of the secondary radiation on images produced by the system.

In a preferred arrangement according to the invention the system further includes: a shutter for selectively allowing and preventing primary radiation from said source to be transmitted towards the subject, a connecting means for selectively connecting the radiation source, shutter, and collimator to the rest of the camera system and a safety interlock which locks the shutter in a closed position when the radiation source, shutter and collimator assembly is removed from the rest of the camera system.

The invention finds particular application where the camera system produces an image from radiation emitted by a radiopharmaceutical material injected into the subject and the transmission radiation source radiation is used to provide data for correction for radiation lost to scattering or attenuation in the subject.

The invention also provides a method of diagnostic imaging in which a subject is injected with a radiopharmaceutical having a first characteristic energy and in which radiation from a transmission radiation source having a second characteristic energy is transmitted from the transmission radiation source through a collimator, through the subject, and to a detector head, part of the radiation from the radiation

source striking internal surfaces of the transmission radiation source and/or collimator causing the emission of secondary radiation, characterised by: limiting the secondary radiation passing through the subject.

One camera system according to the invention and a method in accordance with the invention will now be described, by way of example, with reference to the accompanying drawings in which:-

Figure 1 is a perspective view of the camera system;

Figure 2 is an enlarged end view of a radiation source and collimator assembly of the system with a cover off a radial position adjusting mechanism;

Figure 3 is a perspective view of the radiation source and collimator assembly with a filter slid open to expose a collimator septa;

Figure 4 is a cross-sectional view through the source, shutter, and collimator of the assembly of Figures 2 and 3;

Figure 5 is an exploded view of the radiation source holder; and,

Figure 6 illustrates an interior view along section 6-6 of Figure 5.

With reference to Figure 1, the camera system includes a plurality of gamma camera heads 10a, 10b, 10c, disposed equidistant around a subject examination region 12. A patient couch or other subject support 14 selectively supports a portion of an examined subject in the examination region.

The detector heads are mounted to a rotatable gantry portion 16 which is connected to a suitable motor and bearing assembly supported by a stationary gantry portion 18 to function as a means for rotating or indexing the detector heads around the examination region. Also mounted to the rotating gantry 16 are a plurality of mechanical drives (not shown) for moving each of the gamma camera heads independently radially toward and away from the examination region. The gamma camera heads are preferably mounted on roller carriages or slide bars for smoother, easier movement. The mechanical drives for moving the gamma camera heads radially preferably each include a motor that rotates a screw drive that engages a follower mounted to the gamma camera head.

Each camera head includes a scintillation crystal that responds to incident radiation by producing a flash of light. An array of photomultiplier tubes produce electrical signals in response to each light flash. The signals responsive to the scintillations or flashes of light are combined. The magnitude z of the resultant sum is indicative of the energy of the incident radiation. The relative response of the closest photomultiplier tubes is indicative of the spatial location x, y of the scintillation. An encoder (not shown) indicates the angular orientation θ of the receiving detector head around the examination region.

A transmission radiation source assembly 20 is

also mounted on the rotating gantry portion 16. The transmission radiation source transmits radiation across the examination region 12 to an oppositely disposed detector head 10c. Optionally, additional transmission radiation sources may be disposed opposite others of the detector heads.

A reconstruction processor 30 processes the electrical signals from the detector heads as the detector heads are moved around the examination region to reconstruct a three-dimensional image representation.

Detected radiation from the transmission source is separated from detected radiation from the radiopharmaceutical on the basis of the energy z of the photon peaks. The reconstruction processing means 30 processes the transmission radiation to reconstruct a three-dimensional transmission radiation image representation indicative of the transmission radiation absorptive or blocking properties of the examined region of the subject. The transmission radiation information is used to correct the reconstructed emission radiation image representation in the injected radiopharmaceuticals. The preferred reconstruction and correction processing is illustrated in greater detail in parent U.S. Patent No. 5,210,421.

An operator control panel 32 includes a video monitor 34 for converting selected portions of the emission image representation into a human readable display. Optionally, transmission images might also be displayed. A keyboard 36 enables the operator to control the image reconstruction process, the selection of displayed data, the selection of preselected scanning procedures, and custom operation of the SPECT camera gantry. That is, the operator can control rotation of the rotatable gantry portion 16, movement of the detector heads radially toward and away from the examination region, and positioning of the patient couch 14.

With continuing reference to FIGURE 1 and further reference to FIGURES 2 and 3, the transmission radiation source 20 includes a transmission radiation source holder 40, a transmission radiation source collimator 42, and a means 44 for radially positioning the transmission radiation source. The operator through the console 32 selects a program routine from an acquisition processor 46. The acquisition processor supplies the initial information to a programmable gantry mounted processor 48. More specifically, a motor 50 selectively rotate a drive screw 52 that is received in a threaded drive 54 to which the radiation holder and collimator assembly is mounted. The motor 50 under control from the gantry processor 48 radially positions the transmission radiation source. For stability, the radiation source and collimator assembly are slidably mounted on a fixed rail 56 to insure accurate radial sliding movement. A spatial encoder 58 makes a precise determination of the radial position of the transmission radiation source holder and

collimator assembly. Electromechanical controls controlled through the gantry processor 48 control opening and closing of the shutter for the transmission radiation source and other functions described below.

With continuing reference to FIGURE 2 and further reference to FIGURES 4 and 5, the transmission radiation source holder 40 supports a transmission radiation source 60, e.g. a stainless steel tube about the size of a pencil which is filled with a radioisotope. The radiation source is mounted in a lead shield 62 that has an opening 64 for directing radiation toward the examination region. A shutter assembly 66 is mounted for rotation around the radiation source. The shutter includes an aluminum cylinder or frame 70. The aluminum cylinder supports a radiation blocking or lead circular arc segment 72 which is positioned in front of the opening 64 when the shutter is closed. The aluminum cylinder also supports a tin arc segment 74 which is aligned with the opening 64 when the collimator is rotated 90° clockwise (in FIGURE 4) to a calibration position. The shutter cylinder further includes an opening 76 which is aligned with opening 64 when the cylindrical shutter is rotated counter-clockwise to the open position. The shutter is rotatable between the closed, calibrate, and open position by the electromechanical drive mounted to rotatable gantry portion 16. A manual knob 78 located on the face of the transmission radiation source holder 40 provides a visual indication of the position of the shutter 66.

With particular reference to FIGURES 5, and 6, in order to prevent the operator or a technician from inadvertently opening the shutter when the assembly is not attached to a scanner, a mechanical interlock system 80 is provided. More specifically, the interlock system 80 mounts the transmission radiation source and collimator assembly 40, 42 to the radial positioning means 44 on the rotating gantry portion.

A clamping ring 84 receives a pair of mounting studs 86 from the gantry portion and rotates about 50° to twist and cam lock the assembly 40, 42 to the mounting studs. A pin 88 extends from the clamping ring through an arced aperture 90 to engage a limit switch 92 in the fully locked-on position. The switch 92, thus, provides an electrical feedback to the control panel 32 indicating that the assembly 40, 42 has been locked into place. One or more guide pins 94 assure that the assembly 40, 42 is mounted in the proper orientation.

Once the assembly 40, 42 is locked to the gantry assembly, a shutter coupling clutch assembly 100 connects the shutter 66 with an electromechanical drive 102. The electromechanical drive engages a shutter drive coupling 104 which is connected with an internal drive coupling 106 by a spring biased ball assembly 108. The internal transfer coupling 106 includes a pin 110 that engages a slot 112 in the shutter

66. An end drive sleeve 114 has an arcuate aperture through which the pin 110 extends. The drive end sleeve 114 further has a pin 116 for engaging a slot 118 in the lead line source holder shield 62 to fix the angular orientation of opening 64. A gantry mounted lever 120 selectively lifts the ball 108 against the spring bias decoupling the internal drive member 106 from the shutter drive member 104. This releases the shutter such that it can be manually rotated by the knob 78.

A lockup assembly 130 locks the shutter in the closed position with lead segment 72 across opening 64 unless the assembly 40, 42 is locked to the gantry. The lockup assembly includes a pin 132 which is slidably received in a steel sleeve 134 in the holder assembly. When the shutter 66 is in the closed position, an aperture in a steel guideway 136 in the shutter 66 is aligned with the pin 132. The head of the pin is received in a camway 138 in a shutter locking cam ring 140 which is affixed to the clamping ring 84. (It should be noted that in the exploded view of FIGURE 5, the clamping ring 84 is illustrated positioned on the right side of locking pin 132. When the assembly is completed, the clamping ring 84 is in actuality on the left side of locking pin 132. The locking pin 132 is, in this manner, received between the clamping ring 84 and the shutter locking ring cam 140.) The cam 138 raises radially outward as the clamping ring 84 approaches its closed position to lift the locking pin 132 from the shutter 66, releasing the shutter for rotation.

It will further be noted that in order to rotate the clamping ring 84 to remove the assembly, the cam surface 138 must be permitted to cam the locking pin 132 into the aperture in the shutter. If the shutter is not closed and the aperture is not aligned with the locking pin, then the cam 138 will be unable to cam the locking pin downward and will lock the clamping ring 84 against rotation. Hence, the mechanical interlock 80 cannot be released to remove the line source/collimator assembly 40, 42 from the gantry unless the shutter is in the closed position.

The lockup assembly 130 further includes a pin 142 mounted to the radial drive mechanism 44 fixed to the gantry. The pin extends through an aperture in member 144 to engage a locking element 146. The pin 142 lifts the locking element 146 against the bias of a spring 148 out of a seat in member 144 permitting the clamping ring 84 and locking member 146 to be rotated. To remove the assembly from the gantry, the clamping must be rotated to the position in which (1) the pin 132 is received in guideway 136 and (2) locking element 146 is biased against pin 142. Upon removal, the locking element 146 is seated in member 144 locking the clamping ring against rotation preventing the pin 132 from being lifted.

The shutter 66 has an extending pin 150 which engages through an inner race of a needle bearing 152 which rotatably supports the outer end of the

shutter end 66 to the knob 78. The bearing portion of shutter 66 is notched to connect with a ball and spring detent assembly 154 to a housing portion 156 to provide positioning in the calibrate position. A limit switch 158 is engaged by a groove in the shutter when the shutter is in the calibration position to provide a positive electrical indication thereof.

In order to replace the line source 60, a plug 160 is removed from the line source holder 62. The plug 160 has a keyed end 162 which is adapted to receive a uniquely configured key member. In this manner, access to the radioactive line source is denied to other than a designated key operator.

The radiation line source 60 is preferably filled with a radioisotope that has a half-life, between 6 hrs and 300 days. Because this half-life is much shorter than the life of a SPECT camera, the line source must be accessible to be replaced from time to time.

With reference again to FIGURES 3 and 4, the collimator means 42 includes a pair of lead side walls 200 which diverge at an angle which enables transmission radiation to span the examination region 12. A plurality of thin septa 202 are mounted between the side walls 200. The septa 202 are constructed of a material which has good radiation stopping power and which, when struck by radiation, emits a gamma ray of relatively low energy, e.g. below 50 keV. Preferably, the septa are constructed of tin or an alloy of tin and antimony. Tin emits gamma rays with a characteristic energy of about 30 keV. Other suitable materials for the septa include metals with an atomic number of about 30-70, particularly tin, antimony, zirconium, niobium, molybdenum, germanium, yttrium, cerium, gadolinium, terbium, dysprosium, holmium, erbium, ruthenium, rhodium, palladium, silver, cadmium, indium, tellurium, cesium, barium, and alloys thereof. Of these, tin, antimony, molybdenum, zirconium, and cadmium are preferred for their more ready availability. The side walls of the collimator 200 could also be made of such materials. However, because many scans call for placing the detector heads as closely adjacent as possible, the transmission radiation source and collimator preferably have as narrow a profile as possible. To this end, the greater stopping power of lead, or other high atomic number material, which permits the side walls 200 to be thinner is preferred. It will be noted that the surface area of the septa which are exposed to radiation from the line source is many times greater than the surface area of the side walls. Optionally, the side walls may be covered or plated on their exposed inside surfaces with tin or one of the other above-discussed metals to limit the emission of the 88 keV characteristic x-rays of lead.

A filter 210 extends across an outlet aperture of the collimator 42. The filter includes an inner layer 212 of a material which stops substantially all of the 88 keV energy x-rays from the lead in the side walls of the collimator, yet passes a substantial portion of

the radiation from the transmission line source 60. The filter 210 further includes an outer layer 214 which stops substantially all of the lower energy radiation emitted by the inner layer 212 and by the septa 202. In the preferred embodiment in which the septa are tin or tin/antimony alloy, the inner layer 212 is also tin or tin/antimony and the outer layer 214 is aluminum. The aluminum not only absorbs gamma rays in the 30 keV range, but provide structural strength to the softer tin. Of course, the same alternate metals which can be used to construct the septa can also be used for the inner filter.

Further to the preferred embodiment, the housing has overhanging portions 220 which define a filter receiving track therebetween. A family of filters are provided. The family of filters have varying thicknesses of the inner layer 212. For example, the inner layer 212 in one of the filters may be of an appropriate thickness to absorb about half of the transmission radiation from the radiation source 60. Another of the filters might have a thinner tin portion such that only about 1/4 of the radiation is absorbed. Another one might only absorb 1/10 of the radiation, and so on. As the radiation source decays, the filters are replaced with filters which attenuate a progressively small percent of the radiation. In this manner, the output from the transmission line source assembly can be kept substantially constant over two or more half-lives of the radioisotope in the transmission radiation source. A projection 222 and matching detent 224 in the filters act to snap the filters into position against the natural resiliency of the filter metal. For simplicity of construction and for an assured secure fit, it is preferred that all filters of the family have the same thickness, at least adjacent the edges. Thus, as the tin or more radiation attenuative layer becomes thinner, the aluminum layer becomes thicker.

The invention has been described with reference to the preferred embodiment. Obviously, modifications and alterations will occur to others upon reading and understanding the preceding detailed description. It is intended that the invention be construed as including all such modifications and alterations insofar as they come within the scope of the appended claims or the equivalents thereof.

Claims

1. A nuclear camera system which includes a transmission radiation source (60) which allows primary radiation of a characteristic energy to be transmitted toward a subject, and a collimator (42) for collimating the primary radiation transmitted by the radiation source (60), at least one of the radiation source (60) and collimator (42) tending to emit secondary radiation in response to incident radiation from the radiation source

(60), characterised in that the system includes means for limiting the effect of the secondary radiation on images produced by the system.

2. A camera assembly as set forth in Claim 1 wherein the secondary radiation limiting means comprises material on at least surfaces (200) of the collimator (42), which emits secondary radiation of an energy level that is significantly lower than that of radiation utilised for imaging by the camera system. 5
3. A camera system as set forth in Claim 2 wherein the collimator (42) includes a plurality of thin septa (202) of said material. 10
4. A camera system as set forth in any preceding claim wherein the secondary radiation limiting means comprises a filter (210) disposed between the collimator (42) and the subject incorporating a material (212) for preferentially absorbing the secondary radiation. 15
5. A camera system as set forth in Claim 4 wherein the filter (210) includes a material (212) that absorbs the secondary radiation and reduces the intensity of the transmitted primary radiation and further including a plurality of such filters (210) of different thickness such that as the radiation source (60) diminishes in strength, progressively thinner filters (210) may be substituted to maintain the radiation strength in the subject more nearly constant. 20
6. A camera system as set forth in any one of Claims 2 to 5 wherein said material is selected from the group consisting of: tin, antimony, zirconium, niobium, molybdenum, germanium, yttrium, cerium, gadolinium, terbium, dysprosium, holmium, erbium, ruthenium, rhodium, palladium, silver, cadmium, indium, tellurium, cesium, barium, and alloys thereof. 25
7. A camera system as set forth in any one of Claims 2 to Claim 5 wherein said material comprises tin, antimony, or an alloy thereof. 30
8. A camera system as set forth in any preceding claim further including a shutter (66) for selectively allowing and preventing primary radiation from said source (60) to be transmitted towards the subject. 35
9. A camera system as set forth in Claim 8 further including a connecting means (84, 86) for selectively connecting the radiation source (60), shutter (66), and collimator (42) to the rest of the camera system and a safety interlock (80) which 40

locks the shutter (66) in a closed position when the radiation source (60), shutter (66) and collimator (42) assembly is removed from the rest of the camera system.

10. A camera system as set forth in Claim 9 wherein the safety interlock (80) locks the radiation source (60), shutter (66), and collimator (42) assembly against removal from the rest of the camera system when the shutter (66) is not closed. 45
11. A camera system as set forth in any preceding claim wherein the camera system produces an image from radiation emitted by a radiopharmaceutical material injected into the subject and the transmission radiation source (60) radiation is used to provide data for correction for radiation lost to scattering or attenuation in the subject. 50
12. A method of diagnostic imaging in which a subject is injected with a radiopharmaceutical having a first characteristic energy and in which radiation from a transmission radiation source (60) having a second characteristic energy is transmitted from the transmission radiation source (60) through a collimator (42), through the subject, and to a detector head (10), part of the radiation from the radiation source (60) striking internal surfaces of the transmission radiation source (60) and/or collimator (42) causing the emission of secondary radiation, characterized by: limiting the secondary radiation passing through the subject. 55
13. A method as set forth in Claim 12 wherein the secondary radiation limiting step comprises preferentially absorbing the secondary radiation.
14. A method as set forth in Claim 13 comprising attenuating the transmitted radiation with a filter (210) that also preferentially absorbs the secondary radiation, and, as the radiation source (60) ages, replacing the filter (210) with a filter (210) giving reduced attenuation of the transmitted radiation such that the radiation strength in the subject is more nearly constant over time. 60

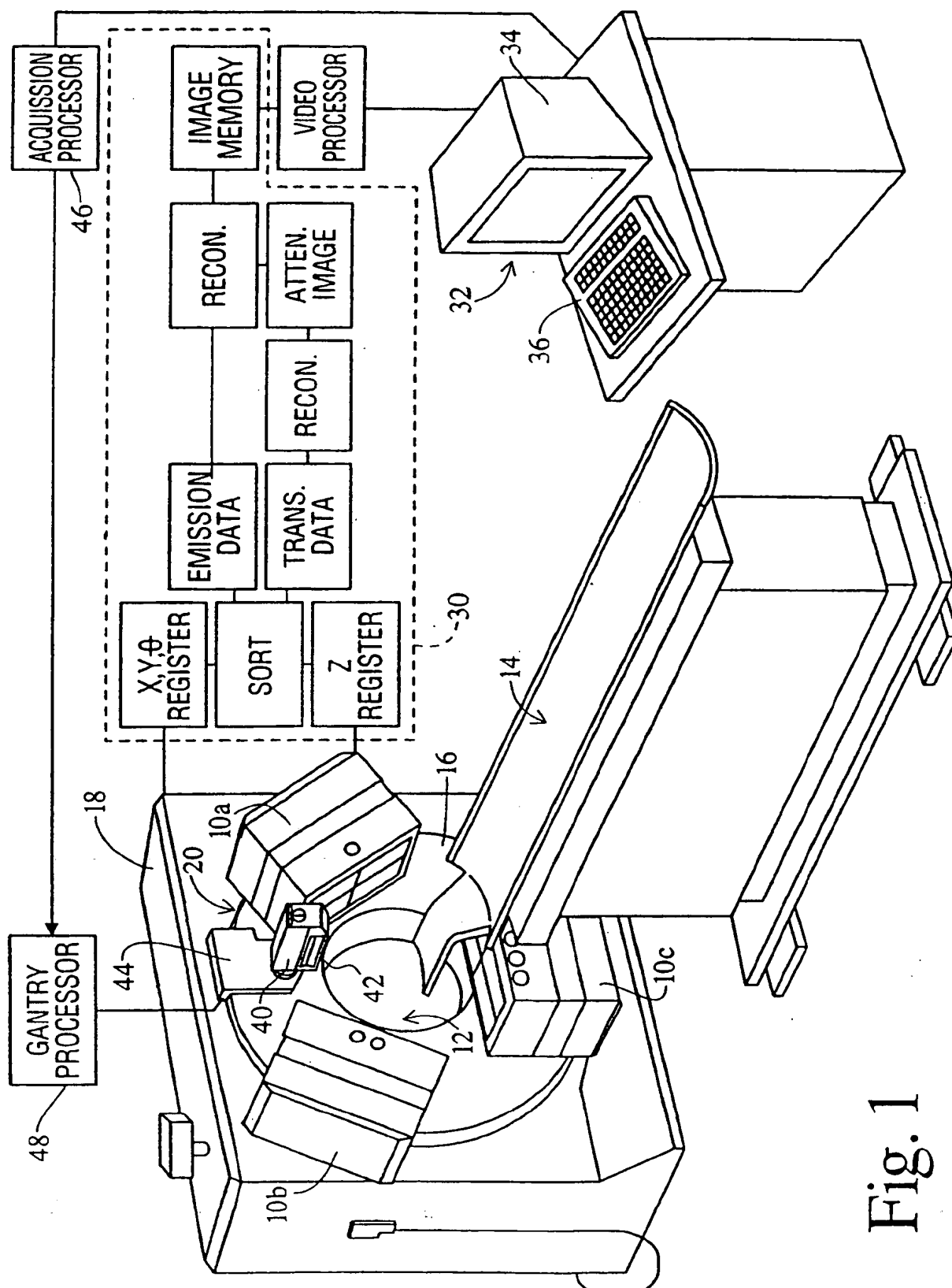


Fig. 1

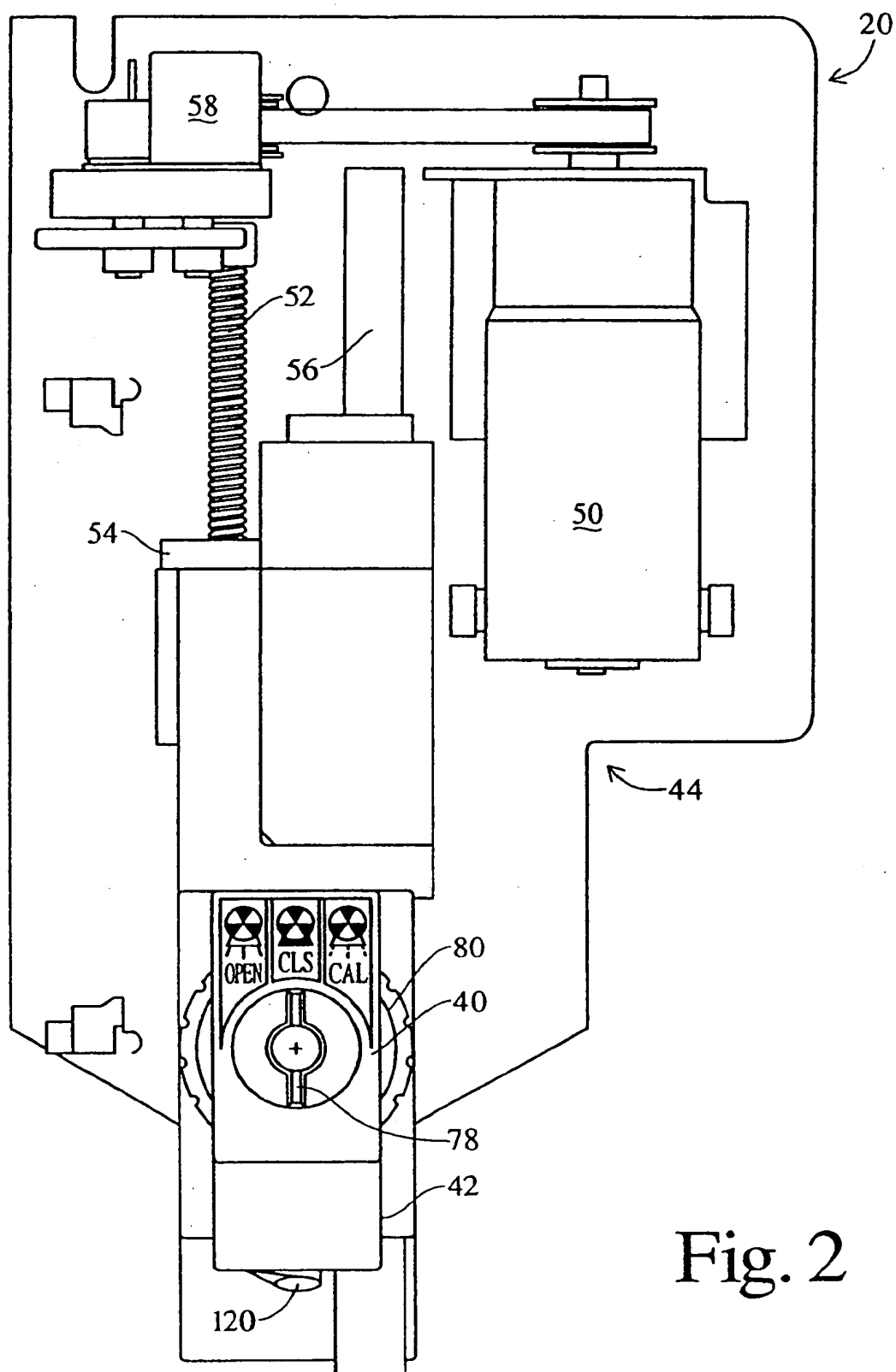


Fig. 2

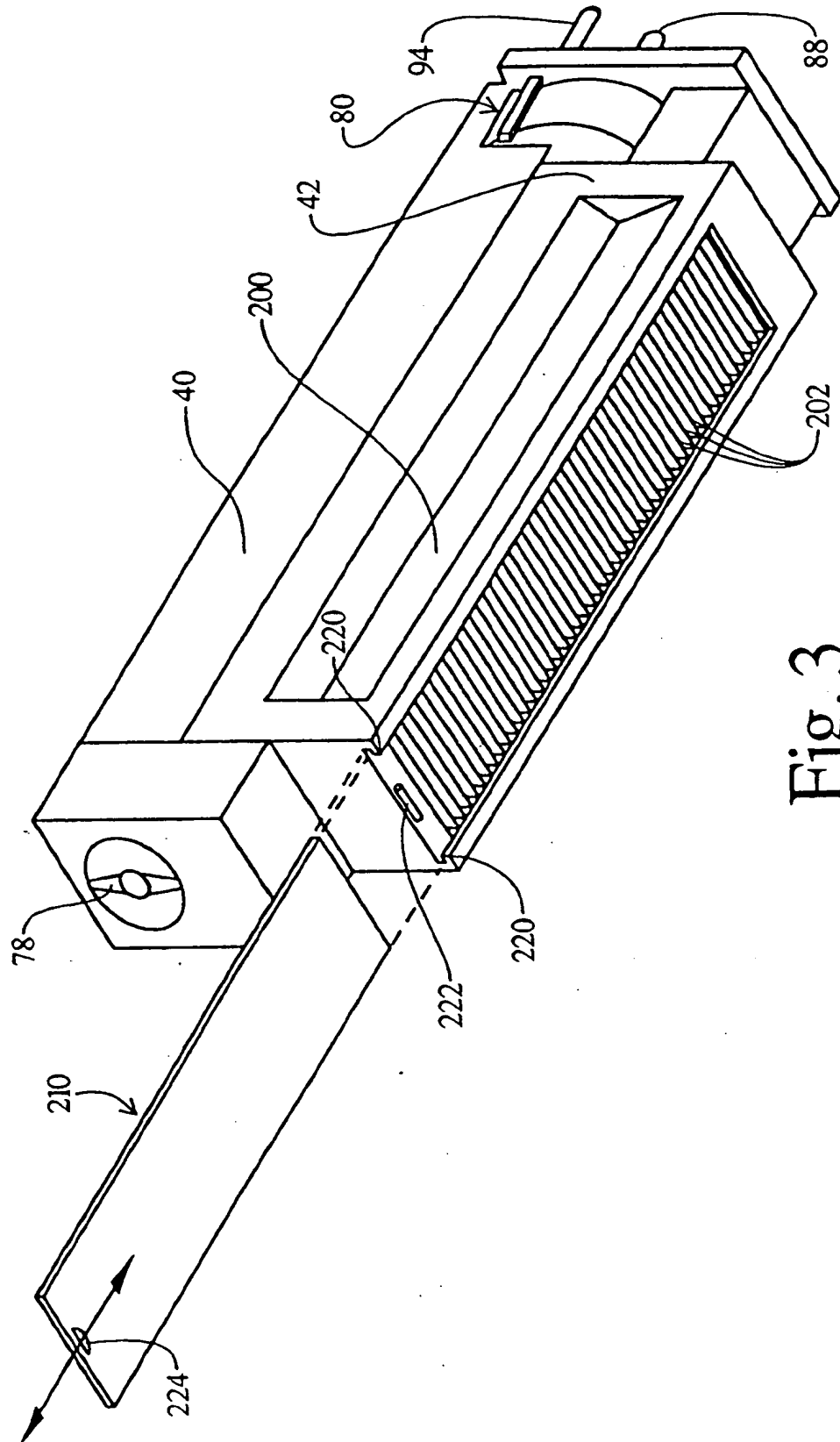


Fig. 3

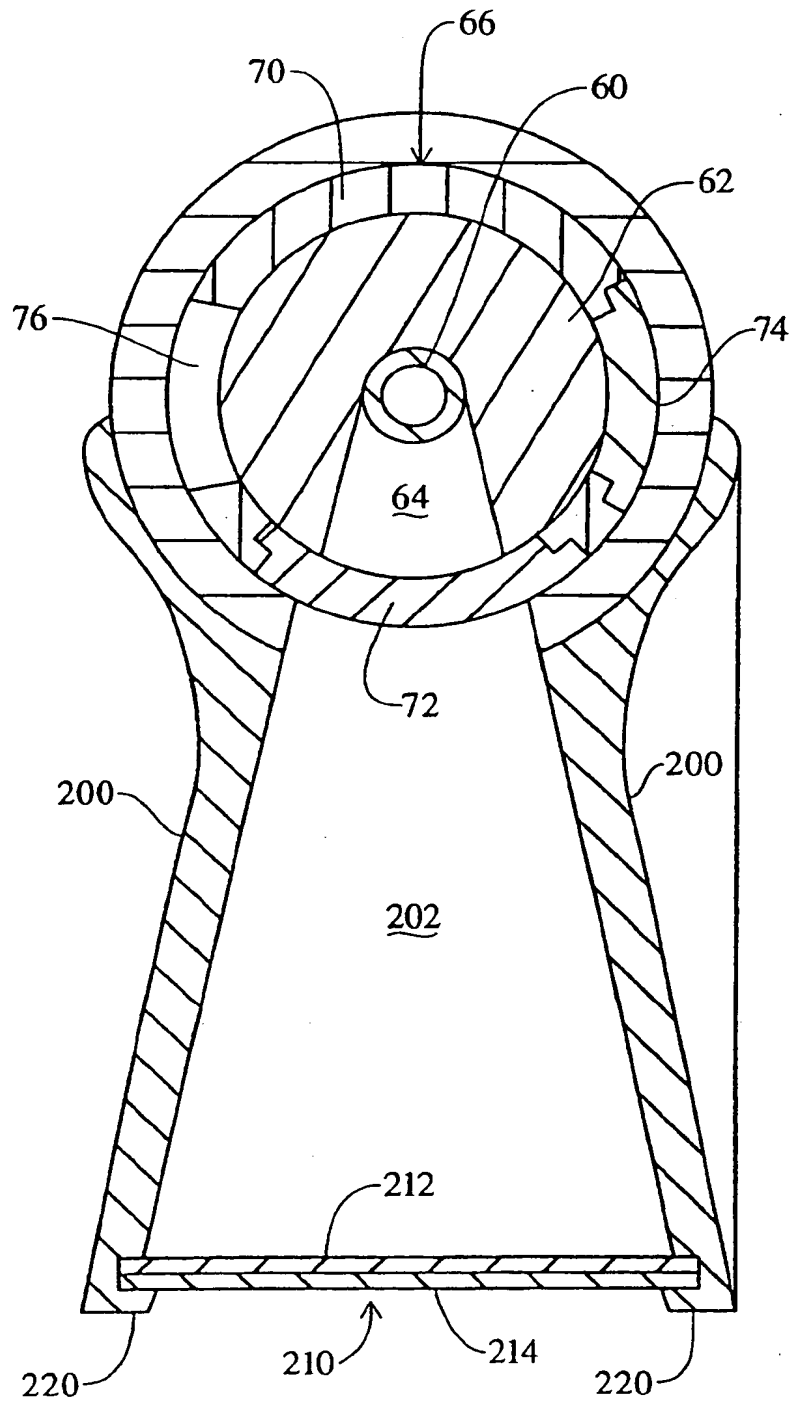


Fig. 4

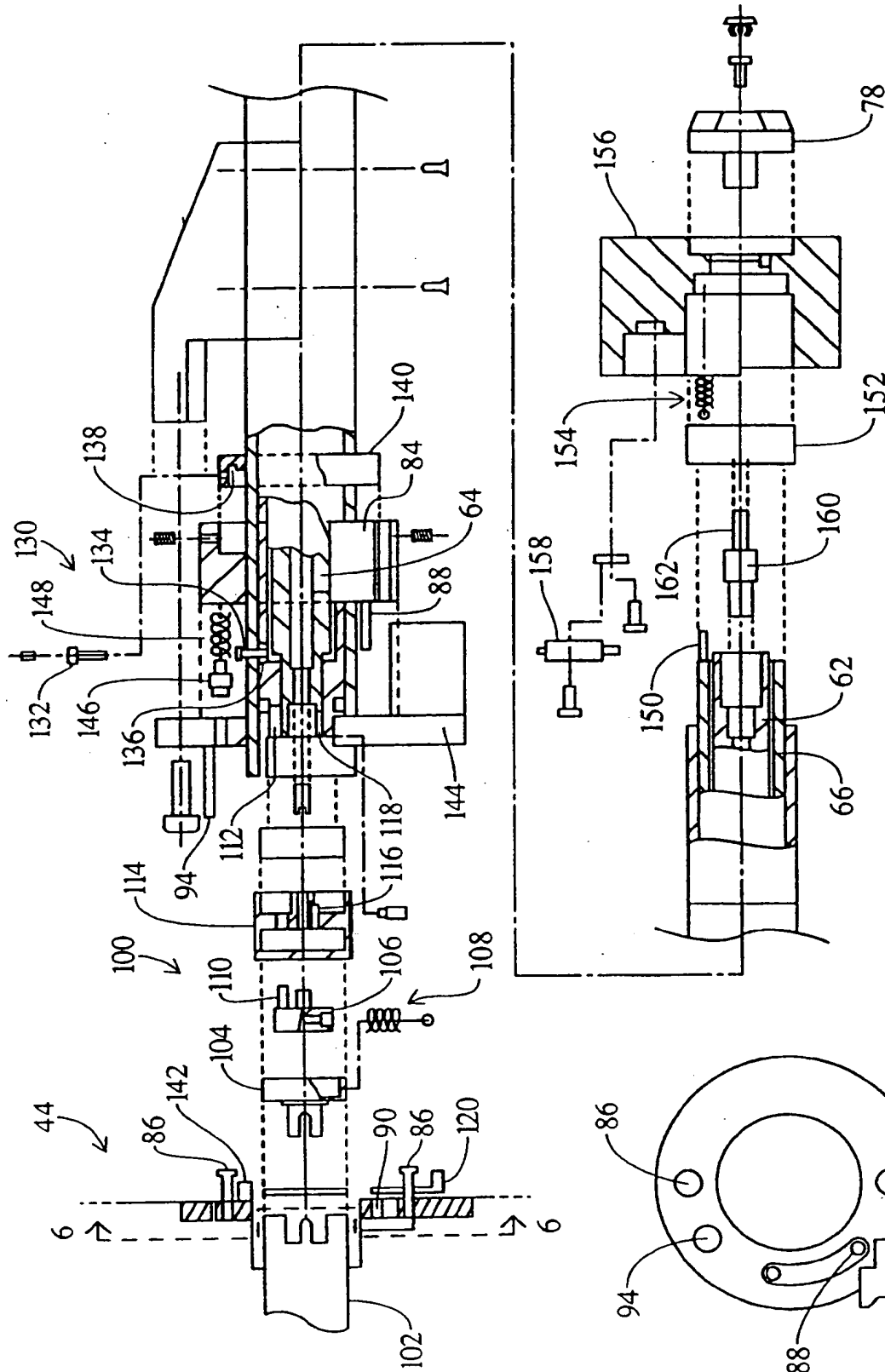


Fig. 5

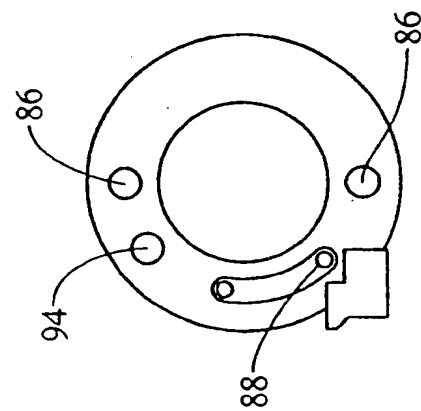


Fig. 6